Invention Disclosure

This form and an annex containing a detailed description of the invention should be forwarded to Mr. Nico van Barschot, Tel. + 31 40 27 44306, Corporate Intellectual Property, building WAH, Prof. Holstlaan 6, 5656 AA Eindhoven, The Netherlands. nico.van.barschot@philips.com.

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Title of the invention

Flexible approach of combining SENSE with moving-bed imaging

Summary of the invention: Please attach a description in English.

Data is acquired while the table is moving relative to the magnet. The whole of the MR-acquisition is split up in a number of "receive-situations". Each of these may be heavily undersampled. During each situation, the excitation- and presaturation profiles "travel" with the patient. Every receive coil "sees" a product of its sensitivity pattern and the combined pattern (excitation, saturation, transmit) of the receive situation. All of these (coils * situations) are used for SENSE-unfolding of a large field-of-view, exceeding the size of the magnet-homogeneity volume.

Description of the invention on annexes; please describe preferred embodiments and their advantages over prior solutions in detail; please include drawings.

Stage and importance of the invention	
a. Stage of the invention?	☐ Research ☒ Pre-development ☐ Development ☐ (trial) manufacture
b. In what products, processes or systems could the invention be used?	MRI equipment
c. For which other business units of Philips could the invention have relevance?	none
d. For which competitors of Philips could the invention have relevance ? Why?	GE, Siemens and others

Distribution of information concerning the invention When, how and where will information concerning the invention be distributed outside Philips? Please consider publications, hearings, exhibitions, offers, contacts with potential customers or suppliers, handing of samples. **N.B. Even after sending this Invention Disclosure to Philips Corporate Intellectual Property, any such acts will impair patentability of the invention. Please contact Philips Corporate Intellectual Property before information concerning the invention leaves Philips.**	

Supplemental information concerning the invention	
a. Is the invention the result of a cooperation with persons outside Philips? If so, with whom?	no
b. Is there, or will there be, an internal report on the invention? If so, please state the number.	yes XJR153-2767
c. Are there, or will there be, other invention disclosures relating to this invention? If so, please state Ref.no.	yes XJS153-2760, XJR-153-2726 (ID 609280), XJR-153-2616 (ID 608652)
d. Are there other persons who could give information on the invention? If so, who?	no
Recommendation management / patent coordinator as to urgency, commercial importance, and competitors' activities.	on list

From: M. Fuderer

To: H. Tuithof, J. Cohen, P. Harvey, J. v.d. Brink
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SENSible approach to generalised moving-bed imaging

Note: contrary to other patent proposals, the document is not split up in distinct sections "introduction" / "prior art" / "proposed improvement". Rather, the full embodiment is described, with a separate section at the end that gives some indication on what is (expected to be) novel and what is prior art.

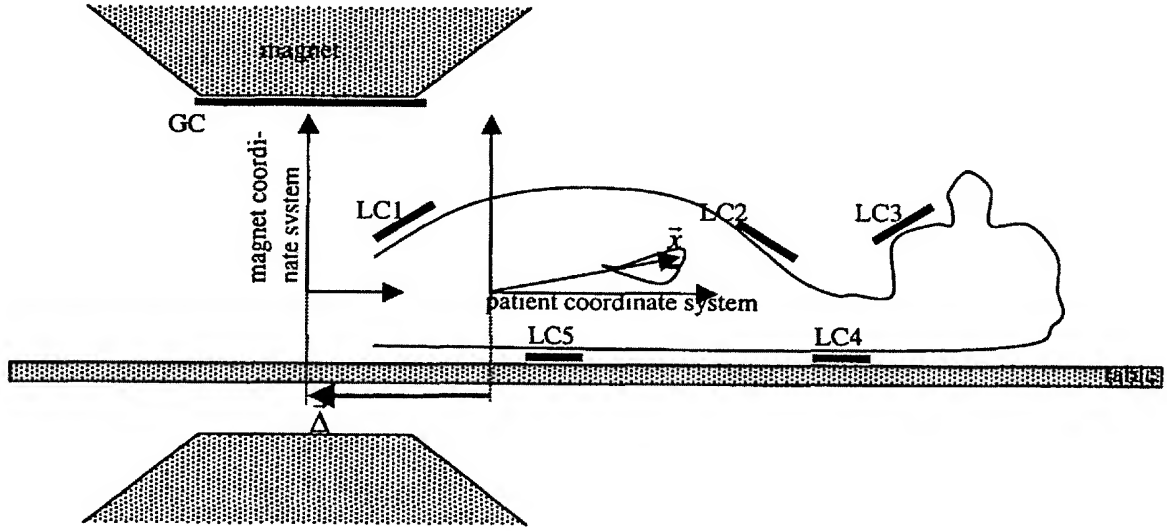
1 Introduction

This document proposes an approach to combine moving-bed-imaging (i.e., acquiring MR data while the bed is moving) with SENSE.
The motivation to use moving-bed-imaging is that the interesting region is larger than the homogeneity volume of the magnet/gradient system.
It is presented in a general way: the approach should be valid (within bounds that are yet to be named/discovered) for any number of MR receive coils, any orientation of the acquired volume and any motion pattern of the patient table.
An essential part of this approach is a SENSE-like reconstruction, but the word SENSE will used in a slightly broader sense here.
An important part of SENSE is coil-sensitivity calibration, which may pose specific problems if the acquisition-region is larger that the homogeneity-region of the magnet. This also has to be acquired during tabletop motion.
Furthermore, reconstructions of moving-table acquisitions will, in general, pose a problem of *geometric mismatch*. This document provides a way to solve this issue.

2 Notions and notations

We have two coordinate systems here, which can translate with respect to each other.
The first one is the *patient*. It is assumed that the patient is "fixed" to the table (or "bed"). One set of coils ("local coils") is supposed to be fixed to the patient. So this set (patient, table, local coils) is fixed to each other. The vector "$\vec{x}$" denotes a position relative to this frame of reference.
The other coordinate system is the *magnet*. Actually, it is the magnet/gradient system. A number of "global receive coils" (generally one or zero) is fixed to this reference system as well as the transmit coil. It is assumed that these coils are large and have smooth sensitivity patterns in the direction(s) of motion.

This coordinate system is offset by $\vec{\Delta}$ relative to the patient system. (Note that this is a "patient-centric" view of the world: the patient is still and the rest of the world, including the magnet, moves around). So a position $\vec{x}$ (relative to the patient) is seen at position $\vec{x}-\vec{\Delta}$ relative to the magnet.



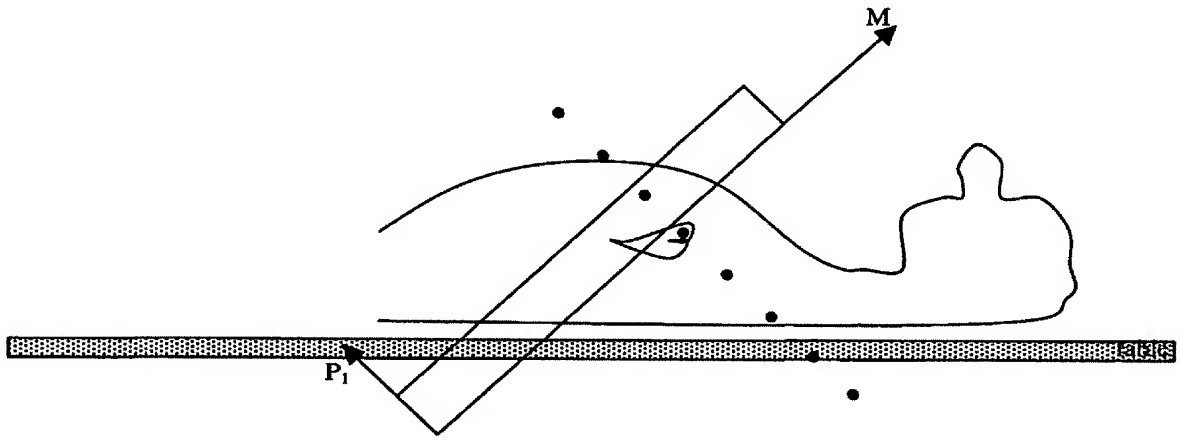
The properties of the patient, e.g., his proton density, are (obviously) given in the patient reference frame; expressed as $\rho(\vec{x})$. Ideally, the same holds for the receive-sensitivity pattern of local coil i: this is $s_i(\vec{x})$. On the other hand, the sensitivity pattern of the global coil j is (ideally, see notes in the sequel) written as $g_j(\vec{x}-\vec{\Delta})$.

Now the notion of "receive situations" is introduced. A "receive situation" is a block of measurements, contiguous in time, during which the table has travelled a relatively small part (e.g., one half or one third or something in between) of the "homogeneity size" of the magnet. During a single receive situation, all magnetisation preparations (like presaturation slabs, volume selections etc.) "travel with the patient". Also during a single receive situation, all of the interesting k-space is covered, except for (potentially significant) undersampling. The full scan covers K receive situations.
In principle, the MR acquisition sequence is identical over all receive situations, except that "magnetisation preparations" may change place and that there may be some offset in the acquired k-space grid (but that is a refinement).
During every receive situation, every coil receives information from the patient. This delivers $(I+J)$ sets of information (some of which may be practically zero, e.g., if a receive coil is completely outside the homogeneity volume). There are K receive situations. So, in total, information has been gathered in $K\cdot(I+J)$ "receive instances".

3 Basic idea

The clinician is interested in a relatively large volume of interest.
Data is acquired while the table is moving relative to the magnet. The displacement thereof relative to the magnet ($\vec{\Delta}$) is known at all times.
The whole of the MR-acquisition is split up in a number of "receive-situations". During each situation, the excitation- and presaturation profiles "travel" with the patient. The acquired data are phase-corrected for the table offset at which they are acquired.

During a receive situation, information from the patient is acquired using any MR imaging sequence of any orientation[1]. Yet, this may be seriously undersampled. An example is sketched.
For every receive situation, we acquire, nominally, the volume indicated by the box. This means that a multitude



of points will fold onto the results of this situation. This multitude may be a 1-dimensional row, a 2-dimensional row or even (like in the case of spiral) a continuum of folding locations.
Every receive-situation has the same subsampling pattern. What differs from situation to situation, is its weighing:

- Excitation- or saturation-profile (with respect to the patient).
- Transmit- pattern of the transmit-coil. (In theory, with a moving table, this also varies *within* a situation – but if the motion per situation is not exceedingly large, this may be neglected).
- Receive-patterns of the "global coils". (Same remark)
- Frequency-response pattern of the receiver system.

In addition, every receive coil (whether local or global) has its own sensitivity pattern.
So every receive instance (= combination of receive coil and receive situation) "sees" a different "overall pattern", which is a product of the coil-sensitivity pattern and the combined pattern (excitation, saturation, transmit and frequency-response) of the receive situation.

The multitude of folding points is unfolded using all the "overall patterns" of every receive-instance. This is done in a SENSE-like manner – although, formally, the unfolding-matrix may be larger. (Nevertheless, the system of equations may be reasonably stable: e.g., for each situation, many folding points may practically "see" a zero pattern, because they fall outside the excitation slab).

[1] The aim to see "any orientation" may seem a bit odd, since we intend to acquire a whole volume anyway. Yet, it may be useful to either orient artefacts (e.g., flow artefacts) in an appropriate direction, or to orient "single slices" to some preferred orientation with respect to some human-tissue structure.

Note that the table motion pattern is flexible: linear, non-linear and even non-monotonic or 2D. The restriction is that it must not travel excessively (e.g., half the homogeneity-volume or less) during one situation.

4 Measuring the patterns

4.1 Introduction

In SENSE, we have the "COCA" measurement, i.e., the reference measurement to establish coil sensitivity profiles.
In this proposal, there is some analogy to COCA, but it is a bit more complex.
(Btw, the general data-acquisition is written for complete freedom of table-motion patterns, including two-dimensional table motion; for this section, a linear one-dimensional movement will be used).

First a review of what is involved:

- At each position of the patient $(\vec{x})$, we have some tissue-properties (all of proton density, T1, T2, flow, diffusion...)
- Also at each position of the patient, we have the sensitivity pattern of the local coils: $S_i(\vec{x})$. Although there might be some dependence on $\vec{\Delta}$ due to coupling to global coils, this is neglected here.
- The pattern of all the coils (global or local, transmit or receive) may be influenced by dielectric resonances, particularly at 3T and up. Here, it is assumed that this can be seen as a purely multiplicative effect on all coil sensitivity – an effect that is only dependent on the patient and the position therein, so not on the coil or coil-position[2]. The dielectric resonance effect is denoted as $D(\vec{x})$. Each coil actually "sees" with sensitivity $D(\vec{x}) \cdot S_i(\vec{x})$.
- The global coils also have their sensitivity pattern, which is approximated as being independent on the actual position of the patient within the magnet frame (except, of course, for dielectric resonance effect, which is explicitly accounted for). This sensitivity is expressed as $g_j(\vec{x}+\vec{\Delta})$.
- The total receive-pattern also may contain effects of receiver frequency response, denoted (given a readout gradient direction) as $F(\vec{x}+\vec{\Delta})$.

For a given measurement type, the *transverse magnetisation* is a non-linear effect depending on *both* the transmit pattern (so on the position within the magnet frame, if a global coil is used for transmit) and the tissue property (so on the position in the patient frame). This is very difficult to handle. An approximation is necessary. That can be achieved by using an FFE measurement with very small tip angle (about 5 degrees or less). In that case, the transverse magnetisation can be approximated as $f(\vec{x}) \cdot \alpha(\vec{x}+\vec{\Delta})$, where f is the "spin density" function, which is a tissue property for a given measurement (in the proposed case, it mainly depends on proton density). α is the tip angle, which is approximated as being proportional to $D(\vec{x}) \cdot g_{\text{transmit}}(\vec{x}+\vec{\Delta})$.

4.2 Problems

For the reconstruction of the proposed method (see other sections), we would ideally like to know all of $S_i(\vec{x})$, $g_j(\vec{x}+\vec{\Delta})$ and $D(\vec{x})$ (but the last one is not crucial). The value of $F(\vec{x}+\vec{\Delta})$ is assumed to be known beforehand, and the value of $f(\vec{x})$ is (at least in principle) not an interesting result of the calibration measurement.

The problem is that we do not measure all of that. What we measure, for a given table position $\vec{\Delta}$, is

- The signal from local coils: $F(\vec{x}+\vec{\Delta}) \cdot S_i(\vec{x}) \cdot D(\vec{x}) \cdot g_{\text{transmit}}(\vec{x}+\vec{\Delta}) \cdot D(\vec{x}) \cdot f(\vec{x})$.

[2] This is an approximation. Yet, a reasonable one: for 3T (128MHz), the effect is mostly monomodal in the head. In the body, the effect is expected to be low because of high damping (high conductivity) of the tissue. For 1.5T, the effect is negligible.

- The signal from global coils: $F(\vec{x}+\vec{\Delta})\cdot g_j(\vec{x}+\vec{\Delta})\cdot D(\vec{x})\cdot g_{\text{transmit}}(\vec{x}+\vec{\Delta})\cdot D(\vec{x})\cdot f(\vec{x})$.

A plain division of those two results in $S_i(\vec{x})/g_j(\vec{x}+\vec{\Delta})$ – which is of no direct use if g_j is not uniform.

4.3 Proposed solution 1

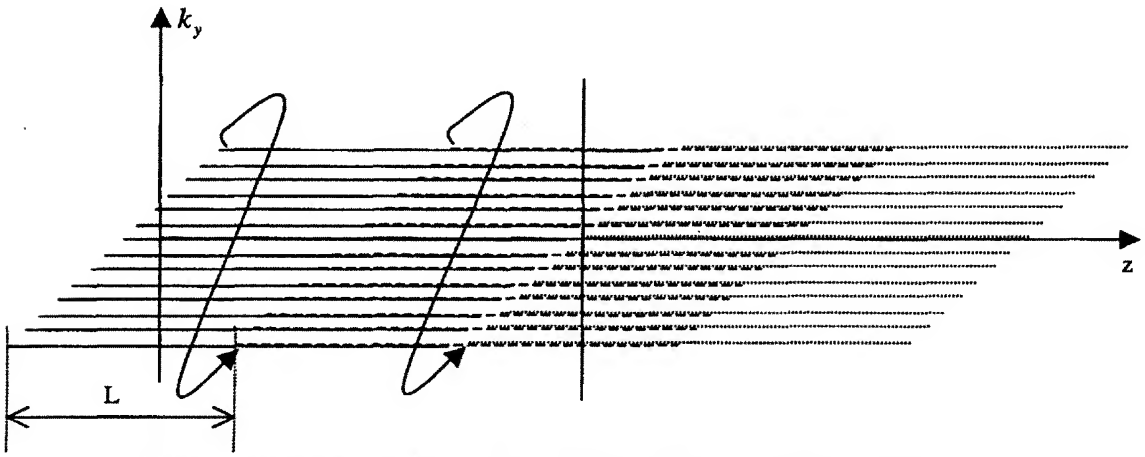
The patterns of the global coils are assumed to be known (well, at least one of them). In practice, that function may be load-dependent, but in our system, this is overcome by the fact that we have a calibrated load-dependance of the sensitivity function, and the load can be easily measured.
A number of discrete reference-scan segments are acquired. During such a segment, the table is still. Each segment results in a full "image". The table is stepped over a number of segments, so for a number of different values of $\vec{\Delta}$.

The knowledge of $g_j(\vec{x}+\vec{\Delta})$ allows calculating $S_i(\vec{x})$. Mathematically, each segment would give the result over the full extent of $\vec{x}$, but in practice, each segment will provide accurate results only over a sub-range of all $\vec{x}$. These (partially overlapping) ranges can be combined using least-squares, resulting in the full map of $S_i(\vec{x})$.

4.4 Proposed solution 2

The (3D) measurement is oriented in such a way that the frequency-encoding is parallel to the table motion direction. We call this the z-direction. The sampling bandwidth is very high in that direction. The other two directions are phase-encodings, one of which is "fast" compared to table motion. For clarity, this is not mentioned further. The "slow" phase-encoding direction is called k_y.

We regard that the magnet is slowly moving with respect to the patient (again, the patient is the centre of the universe). During the motion of the magnet, profiles are acquired with linearly increasing k_y. In the drawing, a line segment indicates the portion of the patient that is "seen" by the homogeneity volume of the magnet.



L is the distance that the magnet covers for a full set of phase-encoding profiles.
The table speed is arranged in such a way that L is maximally half of the homogeneity volume – but less (e.g., ¼) is preferable. Yet, the drawing sketches ½, but this is for clarity thereof.
The consequence is that every profile has been "seen" for at least two different positions of the magnet (except for the outer edge regions, but we disregard these). E.g., profile k_y has been measured for both the displacement Δ_1 and for displacement Δ_2. This allows to approximate the profile that we would have measured if the magnet

were at position Δ_{ref} (as long as this value is in between Δ_1 and Δ_2). This can be done by interpolation (e.g., linear interpolation if two profiles are available, or cubic spline if we have four).
That can be done for every profile, which allows a complete reconstruction of the sensitivity-calibration measurement at magnet-offset Δ_{ref}.

But this can be done for any value of Δ_{ref}. That principle allows acquiring:

- The signal from local coils: $F(\vec{x}+\vec{\Delta})\cdot S_i(\vec{x})\cdot D(\vec{x})\cdot g_{transmit}(\vec{x}+\vec{\Delta})\cdot D(\vec{x})\cdot f(\vec{x})$.
- The signal from global coils: $F(\vec{x}+\vec{\Delta})\cdot g_j(\vec{x}+\vec{\Delta})\cdot D(\vec{x})\cdot g_{transmit}(\vec{x}+\vec{\Delta})\cdot D(\vec{x})\cdot f(\vec{x})$.

But now, contrary to the "problem" section, not for one value of Δ, but for all of them.
This allows to

- separate the signal from local coils into $F(\vec{x}+\vec{\Delta})\cdot g_{transmit}(\vec{x}+\vec{\Delta})$ and $S_i(\vec{x})\cdot D(\vec{x})\cdot D(\vec{x})\cdot f(\vec{x})$,
- and to separate the signal from global coils into $F(\vec{x}+\vec{\Delta})\cdot g_j(\vec{x}+\vec{\Delta})\cdot g_{transmit}(\vec{x}+\vec{\Delta})$ and $D(\vec{x})\cdot D(\vec{x})\cdot f(\vec{x})$

This allows the derivation of $S_i(\vec{x})$ and of $g_j(\vec{x}+\vec{\Delta})$ (Actually, if F is known, then $g_{transmit}(\vec{x}+\vec{\Delta})$ also could be derived, but unfortunately not the dielectric resonance $D(\vec{x})$ – but that is not crucial for the sequel).

5 Receiving and reconstructing the actual data

This has largely been explained in the section "basic idea".
The data is acquired using relatively large steps between profiles, resulting in a (nominally) small "folding volume", i.e., into lots of folding. (For non-Cartesian sequences: relatively sparse sampling). The full extent of k-space is acquired during one scan situation, and the displacement during one scan situation is a fraction (e.g., 1/3) of the homogeneity volume.

Every incoming sample is acquired at a very specific value of $\vec{\Delta}_{sample}$. This is taken into account by multiplying every sample of the incoming data with $\exp(-i\vec{k} \cdot T\vec{\Delta}_{sample})$. Here, T is the coordination transformation from patient coordinates to scan coordinates ("frequency encode"/ "phase encode" frame of reference, which is relevant for oblique scanning). That operation "displaces" the acquired data to the centre of the patient, even if that is way out of the homogeneity volume.

Obviously, one can simplify things by correcting with a fixed $\vec{\Delta}_{profile}$ per profile.

The scan situation has been acquired at an average offset $\vec{\Delta}_{ave}$. This is relevant to estimate:

- The transmit profile for that situation: $g'_{transmit}(\vec{x}) = g_{transmit}(\vec{x} + \vec{\Delta}_{ave})$.
- Similarly for the global-coils receive profile.
- Similarly for the frequency response function for that situation.

The excitation profiles and presaturation profiles are fixed to the patient ("travelling with the table", as seen from the magnet). So these are known in $\vec{x}$.
Also the local coil profiles are known in $\vec{x}$ (see "calibration" section).
All of this can be multiplied into an overall receive pattern for each coil.
In total, if there are K receive situations, then information has been gathered in $K \cdot (I + J)$ receive instances. Each instance has a different overall receive pattern. This allows for a SENSE-like reconstruction, if we have less than $K \cdot (I + J)$ folded points over the entire patient.

In principle, the acquisition stays the same over all receive situations. As a refinement, the phase encoding may be offset by a small amount. This will give some extra phase-encoding, on top of all other mentioned encodings; this will improve the stability of the SENSE-reconstruction.

6 Geometric correction

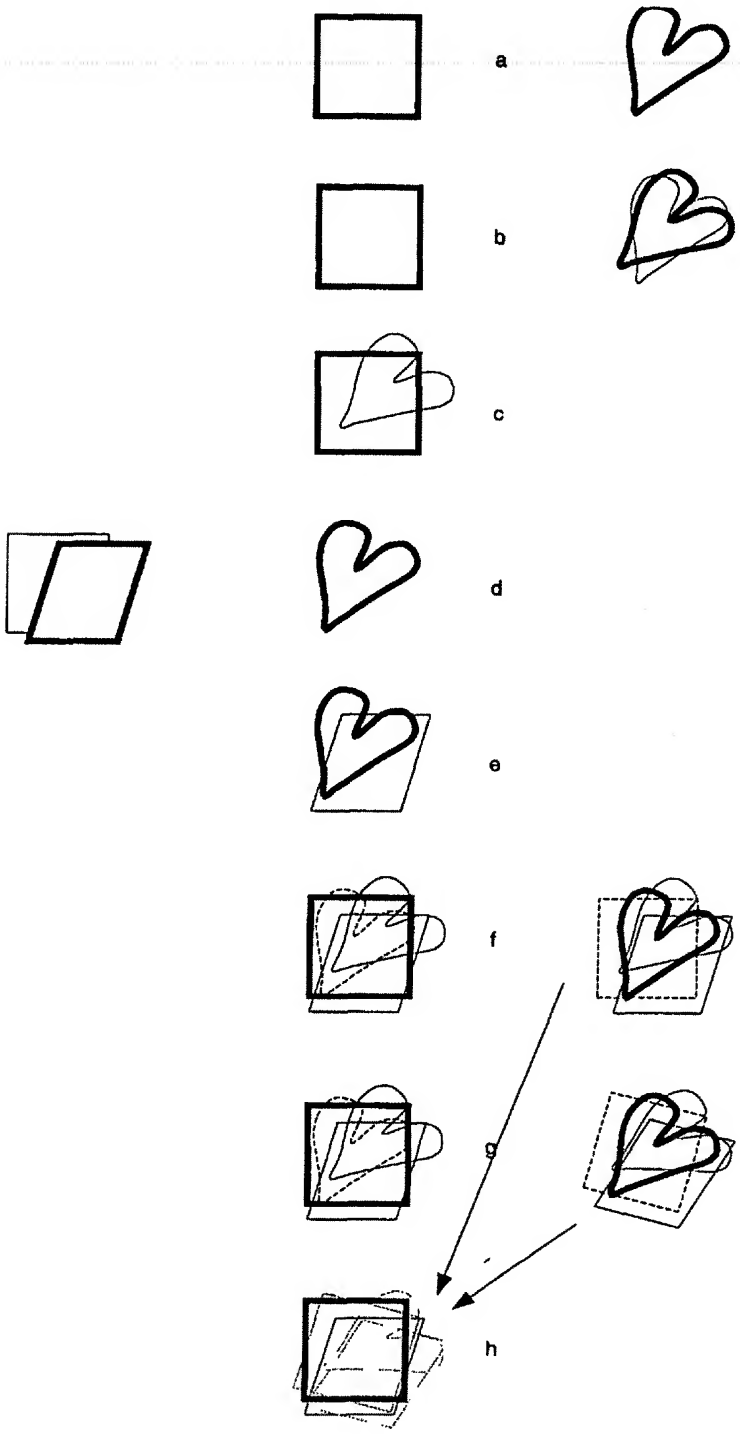
All of above reasoning assumes a perfectly linear gradient system. It assumes that a table-displacement of $\vec{\Delta}$ will cause (a) some different weighting, which has been extensively treated above, and (b) a phase modification of $\exp(i\vec{k}\cdot T\vec{\Delta}_{sample})$ to a sample.

Unfortunately, there is another, significant, complicating factor: geometric distortion. An image acquired at offset $\vec{\Delta}_1$ is geometrically distorted when compared to an image acquired at offset $\vec{\Delta}_2$, even if all weightings (and the displacement $(\vec{\Delta}_2 - \vec{\Delta}_1)$) have been accounted for.

There is a way out.

The principle thereof is that distortions tend to be small where the excitation-and transmit-profiles tend to be large. This can be accomplished (a) by making sequences in such a way that there is not too much excitation in geometrically inhomogeneous, and (b) by designing the system with low transmit-sensitivity in these areas.

The method of the way out is iterative reconstruction. A nice mathematical description could be provided, but here, the explanation is given visually on the next page. The main idea is, in essence, to first reconstruct the resulting image by disregarding the presence of geometric distortion. Of this result, all image regions are distorted in as many ways as there are scan situations. The distorted minus the undistorted versions provide an estimate of "folding distortion-errors". We know the contributions (weights) of every receive-instance to the final image, we can subtract the estimate of the folding distortion-errors from the result.



(a) A simplified patient consists of two parts, a rational one and an emotional one (or: "block" and "heart"). (b) Imaging while the block is largely in magnet-center; the heart is distorted. (c) Due to undersampling, the (distorted) heart folds onto the square; yet, the heart is fainter, since it has seen less excitation. (d) In another receive-situation, the heart is in the center and the square is distorted. (e) Due to undersampling, this is measured in that receive-situation. So (c) and (e) is actually measured. (f) Result after SENSE-unfolding; this consists of two linear combinations of (c) and (e). It is imperfect; e.g., the right part of it consists of the square plus an unsubtracted heart plus a "wrongly subtracted" heart (dashed) and a second-order error (dotted). (g) When (f) is available, we can purposely distort it in the same way that it would have been distorted in (b). (h) Part of (f) can be subtracted from its folded counterpart, and part of (g) can be added, resulting in only second-order artefacts.

7 Novel vs. prior art

- The idea of imaging while moving the table is certainly not new [1-7].
- The proposal to image a number of "scan situations" characterised by a patient-bound excitation profile is likely to be new (yet). Credit to Paul Harvey [7].
- The idea to see multiple stations and to reconstruct them in a SENSE-like manner is suggested by [4] and by [6]. However, the idea to combine local-coil sensitivity profiles with magnet-bound properties (e.g., transmit coil profile) is probably new.
- Likely, the principle is new to see the product of *receive coil number* and *number of receive situations* as independent entries to some SENSE-reconstruction.
- The resulting flexibility (e.g., oblique slice orientation) was formerly unknown.
- The embodiment of acquiring the sensitivity profiles while moving the bed: probably also new.
- Geometric correction principle: likely to be new.

There are many degenerate situations of this proposal that coincide with known methods.

[1] O. Dietrich and J. Hajnal (in ?????)
[2] K. Scheffler, abstract 1774, ISMRM 2001
[3] J. Brittain et al (Stanford/GE), abstract 1726, ISMRM 2000
[4] Y. Zhu (GE), abstract 9, ISMRM 2001
[5] D. Kruger et al, MRM 47:224-231 (2002)
[6] P. Harvey, XJR-153-2726 (ID 609280)
[7] P. Harvey, XJR-153-2616 (ID 608652)